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SAFE HUMAN EXPERIMENTAL EXPOSURE TO IMPACT

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GUIDELINES FOR SAFE HUMAN EXPERIMENTAL EXPOSURE TO IMPACT ACCELERATION

1. Introduction.

The objective of the Naval Biodynamics Laboratory (NBDL) human volunteer impact experiments is to provide a quality data base to be used for modeling human dynamic and physiological responses to impact and to establish the relationship between dynamic response parameters and injury. This information will be used by military and commercial designers of emergency and protective equipment to develop and evaluate improved life-protecting systems for short-term (impact) acceleration exposures. This improved protection should significantly decrease the morbidity and the mortality associated with these exposures.

The human volunteer impact acceleration experiments must be conducted at levels of impact that may produce discomfort but have an acceptablely low probability of producing any permanent or nonreversible injury. Injurious levels of human impact and their correlation with dynamic response parameters must be inferred from experiments with human surrogates or from accident epidemiology data. The candidate human surrogates or analogs are human cadavers, animals with morphology similar to humans, and mathematical models. Each of these surrogates may provide valuable and complementary information concerning injury mechanisms and their correlation with human dynamic response data.

At NBDL, we are conducting animal and human experiments as part of an overall experimental design shown in Figure 1. This figure also shows surrogate data, obtained from non-NBDL sources, together with mathematical modeling efforts and accident epidemiology, as additional inputs to the effort aimed at identifying injurious dynamic response parameters.

2. Planning Human Acceleration Experiments at NBDL.

A. Selecting Peak Sled Acceleration.

Any human experimental impact protocol starts with the least severe level of impact exposure. The impact level is then usually increased during subsequent runs, with the exception of reliability runs* and runs that need to be redone because of faulty data. These reliability and make-up runs may be conducted out of order (i.e., a higher impact level may precede exposure at a lower level). Prior to impact exposure, voluntary motion experiments are conducted in which each subject is requested to pitch, roll and yaw his** head as fast as possible. During these voluntary movements, the same measurements are made as in impact experiments, and the derived kinematic variables allow assessment of the response that can be produced by the subject's muscles.

* A run consists of a single data acquisition event (usually involving exposure to impact) and is assigned a run number. An experiment usually involves many runs. Reliability runs are runs that are repeated with the same subject and impact level to test for within subject measurement reproducibility.

**The masculine term is used because only male subjects participate in impact experiments at NBDL.

This maximum voluntary motion defines the non-stressful response envelope for each subject. For new experimental protocols, the impact levels (i.e, peak sled acceleration) are incremented by only 1 G, beginning with the lowest G-level and ending with the most severe impact. For protocols that replicate previous experiments in which the biodynamic responses were found to be nonstressful, an increment of 2 Gs between impact levels may be employed. For runs which exceed this known nonstressful threshold level, impact levels are again incremented by 1 G. The G-level at which most experimental protocols start is 3 Gs, which is within the voluntary motion range and therefore is known not to be stressful. There are a few exceptional cases in which the lowest level has been 2 Gs. This level was used to evaluate the restraint system and/or the instrumentation. The maximum impact level is selected as the minimal level required to obtain the necessary data (see section 2.B. below).

B. Evaluation of Experimental Severity and Risk.

The single most important consideration in these experiments is to conduct the required human impact runs in a manner which minimizes the risk of any permanent or nonreversible injury to the subject, while at the same time providing valid information required to meet program objectives. In order to obtain valid and meaningful data, some acceptable risk of injury to the human subjects must be defined. The intention at NBDL is not to expose human volunteers to impact levels that have high risk of producing injuries or that compromise the future health and well-being of the subjects. This policy contrasts with operational conditions, where tolerance levels for crash survival may permit injury levels which are not acceptable for human experiments conducted in a laboratory. Examples of unacceptable injuries are any fractures of bones, dislocation of vertebral bodies, subluxation of vertebral bodies, herniation of discs or fracture of disc endplates, avulsion of ligaments, disruption of blood vessels, internal viscera or supporting ligaments that result in any chronic impairment of health, and any functional injury. However, experiments may produce minor injuries such as muscle soreness, nonchronic tissue strains, external fascia abrasions and contusions due to restraint interfaces with the subject, short-duration mild headaches, brief periods of brachycardia (i.e., a few complexes) or tachycardia occurring immediately after the impact exposure but which progressively return to normal rates, other anomalies or arrythmias in the EKG complex not considered medically significant and which return to normal in a brief period, and brief "stunning" or mild concussion similar to that observed in sports activities and which are free of residual medical effects.

The remainder of this section presents tolerance levels and severity indices that have been correlated with injury and explains how the scientific and medical staff at NBDL use these data to determine the maximum acceptable impact severity for an experimental protocol.

The severity of an abrupt acceleration exposure is a complicated function of the duration and magnitude of the acceleration profile, the direction of the acceleration vector relative to the anatomical axes, and the restraint of the human subject. For simple, unimodal acceleration profiles, severity is a function of one or more of the following variables:

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(1) Direction of acceleration relative to the anatomy;

(2) Peak acceleration level of profile;

(3) Rate of onset of the acceleration profile (how rapidly it rises to peak acceleration);

(4) Duration or dwell (the length of time the acceleration remains above a high percentage of the peak profile acceleration);

(5) Rate of offset (how rapidly the acceleration profile drops off after the peak acceleration level);

(6) The posture and manner in which the subject is restrained; and,

(7) The static and dynamic tension of the restraint prior to impact.

The use of these variables to define tolerance levels requires that the human subject be restrained in a consistent and repeatable fashion.

An alternate definition of tolerance levels evaluates the response of a mathematical model to the input acceleration and expresses the severity and tolerance levels in terms of response variables of the model (i.e., Dynamic Response Indicator, DRI). Other tolerance levels may be expressed in terms of measured response parameters on an analog model (i.e., manikin), or more directly expressed in terms of severity indicators derived from measured response variables on the human.

Regardless of the definition of severity or level of acceleration, acceptable levels of exposure are obtained from one or more of the following sources.

(1) Exposures employed in previous research with human subjects.

If it can be demonstrated that the severity of the planned protocol is less or no more severe than the results from an existing NBDL experimental database, then the medical and injury-related experience from this database is considered sufficient to justify the risk involved in the recommended exposure level. The most severe exposures conducted at NBDL for each acceleration vector are presented in Table 1.

If human experimental data from other laboratories are available, and the exposures, restraints, number of subjects, and related injury experiences are sufficiently documented, then these data can be used to justify the maximum exposures recommended in the NBDL protocol, provided that the recommended NBDL exposures are less severe than those documented exposures at which unacceptable injuries occurred.

(2) Human exposure data from non-experimental sources.

If injury statistics are available from accident epidemiology or operational events (e.g., aircraft ejections) for which exposure and injury data are suitably defined, then these data can serve as sources for defining acceptable levels of human exposure in an NBDL experimental protocol.

(3) Work conducted on human surrogates/human cadavers.

If available animal/cadaver injury data can be scaled to human tolerance levels for injury in a rational manner and verified by comparison with human accident data, then these data may also serve as sources of useful information in defining acceptable levels of exposure for NBDL human impact experiments.

(4) Mathematical modeling techniques.

Available statistical regression models can be used to extrapolate exposure limits from existing experimental data and these extrapolated limits can be compared with other sources of information (e.g., accident epidemiology). Validated mathematical models also can be used to predict dynamic response parameters for experimental conditions not yet initiated (i.e., combinations of vectors, restraint perturbations, and acceleration profile perturbations).

If, in addition to the four sources described above, published human tolerance levels for non-injurious exposures to acceleration are to be used, it is important to know how these recommended levels were generated, as well as their limitations. The biodynamic effects of deviations in the planned experiment from the conditions under which these tolerance levels were determined originally, must be accounted for in the selection of the maximum exposure level.

It is noteworthy that although an experimental protocol may recommend a given maximum exposure level, the on-site experimental monitoring of critical dynamic and physiological variables or severity indicators (e.g., EKG, evoked potentials) may dictate that the experiment be halted at some lower level than that originally planned.

Guidelines for determining safe maximum levels of impact exposure, based on the above sources, are presented in Section 3.

3. Guidelines to the Committee for the Protection of Human Subjects (CPHS).

The reference section (pages 26-28) includes references to impact data obtained at NBDL and elsewhere, compilations of recommended tolerance levels, and cadaver data that can be used by the CPHS to verify and judge independently the maximum exposures recommended in an experimental protocol. These references, together with the Committee's collective knowledge of the experimental area and sound judgement, are the basis for evaluating risk of injury. The following discussion reviews the information form source listed in the reference section. These sources are referenced in the text by parenthetical numbers corresponding to the number provided in the reference section. This review is limited to the body segments at highest risk in current NBDL impact experiments (i.e., the head, cervical spine, spine other than cervical, thorax, pelvis, and internal viscera).

A. Data from previous human research.

As stated previously, experimental data collected from past experiments at NBDL, together with observed medical effects, provide a basis for recommending other experiments which do not exceed these previous impact levels. Tables 1-4 and Figures 2-5 present summaries of the results from the most severe runs conducted at NBDL. Table 1 lists the parameters describing the sled acceleration profiles for these runs. The number of subjects, initial conditions, and restraints are also defined in this table. Table 2 presents parameters related to injury tolerance levels derived from measurements made on human body segments during these runs. Torques and forces at the occipital condyles in the head anatomical coordinate system and head injury criteria (HIC) are shown for cases involving the most severe G levels used at NBDL.

Plots of the torque and force components and resultant magnitudes, together with the HIC, are presented for a -X exposure in Figures 2-5. These same plots are available or can be readily obtained for any acceleration exposure in the NBDL data base. Reference (9) recommends tolerance levels defined in terms of these injury-related parameters, as well as the equivalent torque (moment) at the occipital condyles, and the forces at the condyles, for use in assessing the risk of neck injury (i.e., injury to the cervical spine). This reference also discusses the rationale for selecting the Gadd Severity Index (GSI) or the HIC as indicators of risk of concussion during direct impact.

Table 3 was constructed using NBDL data (6), and shows the peak angular acceleration, angular velocity, and linear acceleration for representative NBDL maximum severity exposures in the -X and +Y directions. These data are used to define tolerance levels for concussion based on angular acceleration and angular velocity (i.e., shear failures). In addition, the linear relationship between intercranial pressure and peak acceleration (7) suggests that acceleration and duration may be better indicators of injury than the GSI or HIC since these latter indicators are computed using acceleration raised to a power.

The peak seat pan loading and the belt loads for the +Z runs at NBDL are shown in Table 4 for the most severe exposures in this direction. These data can be used to evaluate potential injury to the pelvis, spine and thorax for runs in this direction.

The medical findings in the NBDL impact exposure data can be summarized as follows.

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(1) Several instances of fainting, resulting from restraint and psychological stress, have occurred before and after impact. These fainting episodes were not correlated with the level of impact exposure (i.e., the episodes varied randomly across impact levels). One case of restraint-induced cardiac asystole for a duration of 8 to 9 seconds occured just prior to impact (i.e., the asystole occured prior to activation of the impact system). In all these cases, the run was immediately aborted, the subject removed from the sled, and immediate medical attention was provided. Recovery was uneventful in all instances.

(2) Headache is the most common finding immediately after impact, and is related to exposure severity in both the -X and +Y directions. The headaches can be severe for up to one minute, and gradually diminish with no sequalae.

(3) It was common to have muscular myalgia on the side of the neck opposite head motion for exposure in the +Y direction. This condition is related to the maximum head angle relative to the neck angle, and can be controlled by limiting the endstroke velocity of the sled as the G-level is increased. These neck problems have not been observed in the -X direction. There was a case in the -X+Y direction in which a subject had a sore neck that persisted for two weeks. The subject recovered but was disqualified from the program. Subsequent medical examinations as part of the long-term follow-up program* revealed no sequalae in this subject.

(4) Several cardiac findings are summarized in reference (34) for the -X and +Y directions. For the most part, these findings were medically insignificant. One serious cardiac finding occurred in one subject run in the +Z direction, and was characterized by two premature ventricular contraction (PVC) complexes in succession, followed by a normal complex, and then followed by another PVC prior to returning to normal. This condition can be a forerunner of ventricular tachycardia and was considered potentially dangerous. This subject was disqualified from the program with no sequalae. It is interesting to note that other researchers have produced this same type of anomaly during centrifuge exposures (28).

(5) One subject had pain radiating to his left arm after a +Y exposure. This condition was diagnosed as a stretched brachial plexus and the subject recovered with no sequalae.

(6) One subject was disqualified from the program after a +Z exposure that resulted in pain in the ischium. This subject recovered with no sequalae.

These medical findings are generally descriptive of the type of medical events observed. This information exists for each run in the NBDL data base and can be made available to the CPHS to support risk assessments of proposed experimental protocols.

* The subject is brought back after a period of three years and given the same medical examination that was given at entrance and discharge from the program.

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Some additional +Y impact data from non-NBDL experiments are available (27). In most of these experiments, the head and neck were restrained and the subjects were exposed to high G-levels (20-30 Gs) without injury. In one series of experiments, human subjects were exposed to +Y impact levels of 12 G with upper torso restraint and to 9 Gs without upper torso restraint. Head and neck restraints were not used for either of these exposures. Although no permanent physiological effects were noted in these experiments, physical complaints (such as neck stiffness) occurred after most runs above 6 Gs. Therefore, with the torso restrained, head angular deflection should be considered potentially hazardous from a medical standpoint.

For the unrestrained head and neck, the data collected at NBDL best describe the allowable tolerance levels for the +Y direction. For the noncervical spine in a well-restrained human subject, these tolerance levels are much lower than the levels at which injury may occur.

B. Tolerance levels - Head.

Many injury criteria have been developed which relate the acceleration profile of the head to survivable head injury. With regard to the effects of linear acceleration, most of these criteria were originally derived from the Wayne State Tolerance Curve (WSTC), published in 1962 (35), and reproduced in Figure 8. The WSTC uses linear skull fracture as the criterion for injury and as such is of limited value in assessing the effect of indirect impact to the head as encountered in the experiments at NBDL. The direct impact injury assessment is based on average acceleration and pulse duration. The maximum risk defined by the area below the WSTC is cerebral concussion without permanent aftereffects, while the area above the WSTC is considered to be potentially hazardous and life-threatening. An excellent discussion describing the supporting model and definitions of many of these head injury-related criteria is available (8). Two commonly used criteria, the Gadd Severity Index (GSI) and the Head Injury Criterion (HIC) (detailed in (9)) are presented in these guidelines to evaluate the medical effects of linear acceleration. A discussion of the origin and important limitations of these particular indices also is available (10, 11). These criteria for head injury are presented below in terms of index (GSI or HIC), index tolerance level, and references.

(1) Linear accelerations.

(a) Direct impact

Index	Tolerance level	Reference
GSI	< 1000	(9)
HIC	< 1000	(9)

(b) Indirect impact

Index	Tolerance level	Reference
GSI	< 1500	(12)

(c) Reference (9) notes that an HIC greater than 1000 only implies a concussion hazard for pulse widths (durations) of less than 15 msec.

Angular acceleration and velocity have been incorporated in a model of head injury based on shear stress (13,14). The tolerance criteria developed from this model were supported by rhesus and squirrel monkey experiments and scaled to humans using brain mass ratios and the shear failure model. Limiting values were determined for angular acceleration and velocity which predicted a 50% probability of concussion (without direct impact to the head) in humans having a brain mass of 1.3 kg. Though the scaling of the model to humans has not been validated, these values suggest possible tolerance levels.

(2) Angular acceleration and velocity.

Index	Tolerance level	Reference
Angular acceleration	< 1800 rad/sec^2	(15,16)
Angular velocity	< 50 rad/sec	(15,16)

In the sled runs at NBDL, the maximum HIC number observed is less than 200 (see Table 2), and it is doubtful that these experiments would result in an HIC number representing a potential hazard without first having serious neck muscle problems or a severe chin strike on the chest. The usefulness of these criteria at NBDL is for impact in which the head returns to the headrest (e.g., +Z exposures using the horizontal sled). A more pertinent concern with regard to concussion in the NBDL experiments involves exceeding angular acceleration tolerance levels. Angular velocity can be controlled by changing the sled maximum or endstroke velocity. Angular acceleration, however, is controlled mainly by peak sled acceleration and can only be significantly varied by changing the initial head and neck position.

From Table 3, the NBDL average maximum peak angular acceleration for the -15 Gx runs was 1800 rad/sec^2, which is the level projected for 50% probability for concussion. We do not intend to exceed this G-level or angular acceleration until we have convincing evidence from our own rhesus experiments that it is safe to exceed these levels. The scaling solution from rhesus to humans is particularly straightforward in this situation since the tolerance level was derived using rhesus data (13,14). Colonel Stapp's run on the rocket sled in excess of 40 Gs would indicate that the limit of -15 Gx is extremely conservative.

The tolerance levels derived for the head were determined from anterior-posterior direct impact and X indirect impact experiments. For the Y direction, the shear model and tolerance levels may be different. NBDL data on the unrestrained head and neck indicate that neck-related problems precede head injury, and therefore neck-related problems would be the limiting factor in defining maximum impact exposures.

C. Tolerance levels - Cervical Spine and Neck.

The human head has a mass of approximately 4.5 kg with an internal brain mass of approximately 1.3 kg. The occipital condyles at the base of the skull located on each side of the foramen magnum rest on the superior facets of the atlas, with the head load transmitted through the cervical spine, thoracic spine, and lumbar spine to the pelvis.

The tolerance limits for the neck are provided in reference (9), and are expressed in terms of the equivalent torque at the occipital condyles and the shear and axial forces at the condyles. The equivalent torque at the condyles is the torque that the neck produces on the head that is consistent with the mass distribution properties of the head (mass, moment of inertia tensor, center of gravity location), the location of the head/neck joint (condyles), and the observed kinematic motion of the head. Unfortunately, the equivalent torque is sensitive to the location of the head center of gravity relative to the condyles, and this location should be defined as well as the corresponding torque thresholds. The force components are robust and not sensitive to geometric configuration factors. The recommended tolerance levels come from static tests on living human volunteers, and dynamic tests on human volunteers and human cadavers.

These tolerance levels were originally discussed in reference (17), and the recommendations in reference (9) are consistent with this original reference and are summarized below.

(1) Neck flexion.

(a) Maximum equivalent moment and neck shear force from living human dynamic experiments.

Parameter	Tolerance Level	Reference
Equivalent moment	< 88.2 Newton-Meters (N-M)	(9)
Shear force at condyles	< 787 Newtons (N)	(9)

(b) Maximum equivalent moment and shear force determined from dynamic cadaver experiments without producing ligamentous or bone damage. The levels are based on cadaver responses with the chin in contact with the chest.

Parameter	Tolerance Level	Reference
Equivalent moment	< 192 N-M	(9)
Shear force at condyles	< 1944 N	(9)

(2) Neck extension.

(a) Tolerance levels determined from noninjurious living human dynamic tests.

Parameter	Tolerance Level	Reference
Equivalent moment at condyles	< 30.5 N-M	(9)
Neck shear force at condyles	< 231 N	(9)
Neck axial force	< 249 N	(9)

(b) Tolerance levels determined from dynamic tests on human cadavers.

Parameter	Tolerance Level	Reference
Equivalent moment at condyles	< 47 N-M	(9)

It is important to note that reference (9) does not give accelerations, but reference (32) indicates that a 6 G peak acceleration of .12 sec. duration may be the limit of voluntary human tolerance for unrestrained neck extension.

(3) Lateral flexion.

(a) Tolerance determined from dynamic tests on living human volunteers.

Parameter	Tolerance Level	Reference
Equivalent moment at condyles	< 45.2 N-M	(9)
Shear force at condyles	< 792 N	(9)

(4) Neck axial and transverse tolerance levels.

(a) Shear force and axial forces determined from static tests on living human volunteers.

Parameter	Tolerance Level	Reference
Shear forces:		
Anterior-posterior	< 845 N	(9)
Posterior-anterior	< 845 N	(9)
Left to right/right to left	< 400 N	(9)
Axial forces:		
Tension	< 1134 N	(9)
Compression	< 1112 N	(9)

(b) Axial failure loads determined from dynamic tests on intact human cadaver cervical spines.

Parameter	Tolerance Level	Reference
Tension	< 2450 N	(18)

(c) Compression failures on isolated human cadaver cervical spines using dynamic tests.

Parameter	Tolerance Level	Reference
Compression	< 1779 N	(18)

The tolerance levels for living human voluntees are defined and developed only for minimum risk injury. The tolerance levels developed from human cadaver experiments do not exclude the possibility of muscle damage at a lower acceleration level.

D. Tolerance levels - Whole Body.

For the purpose of these guidelines, the torso is considered to be all the elements of the human anatomy excluding the limbs and the head and neck, and represents approximately 60% of body weight. The torso is important to impact experimentation because it contains and protects many of the vital organs of the body. For a seated and restrained subject, exterior surfaces of the torso are often the places where external forces are first coupled to the body. The major load-bearing element of the torso is the bony skeleton (pelvis, spinal column (sacrum-T1)), the bony rib cage, and the shoulder girdle (scapula and clavicle). The elements of the bony, skeletal structure are rigid bodies that articulate with each other, allowing considerable changes in the body configuration. The transmission of forces between the elements of the bony structure of the torso is accomplished through a profuse system of soft connective tissue and fascia (ligaments, discs, and muscles). External loadings of torso elements are transmitted through the skin and external fascia of the torso, and the interface or joint articulations and connective tissue between the neck and torso and the between the limbs and torso. The bony structure of the torso protects critical elements of the viscera. These elements interact dynamically with the soft tissue and bony structure of the torso, thereby changing the loading on elements of the torso and changing the mass distribution of the torso.

In general, the greater the area of the torso over which the load is distributed, the greater the tolerance level. In this section, whole-body (torso) tolerance levels are presented for -X, +X, Y, ±Z direction accelerations. It should be noted that the head and neck are assumed <u>constrained</u> at these tolerance levels.

In the experiments at NBDL, loads are transmitted to a seated subject from the acceleration device seat through the seat back and the seat pan, and the shoulder and lap belts to the surfaces and anatomical structures of the torso. For exposures in the -X direction, loads are mainly transmitted through the shoulder and lap belts to the anterior thorax and pelvis, respectively. For the +X direction, loads are transmitted through the seat back to the thoracic spine and shoulder complex, posterior pelvis, and lumbar spine. For the +Z direction, loads are transmitted from the seat pan through the fatty tissue of the buttocks to the ischial tuberosities of the pelvis and through the sacrum and spinal vertebral bodies, and to a lesser extent through the shoulder belts to the shoulder girdle and anterior surface of the thorax. In the -Z direction, loads are transmitted through belts on the femur and lap to the pelvis, and then by way of the sacrum to the rest of the spine. If shoulder belt restraint is used during -Z exposures, forces may be transmitted to the subject through the shoulder girdle complex and thorax. In the +Y direction, forces are mainly transmitted to the thorax through the shoulder girdle (which is in contact with a sideboard restraint). The shoulder belts and lap belt may also transmit loads to the subject during +Y exposures.

The Federal Motor Vehicle Safety Standard (FMVSS 208; reference (9)) specifies as acceptable any anterior-posterior (A-P) acceleration pulse which does not exceed 60 Gs for a cumulative time period of more than 3 msec. Previously, FMVSS 208 had specified a GSI of less than 1000 as acceptable. Acceleration is measured at the center of gravity of the dummy thorax (presumably, part 572). Reference (19) reports measurements made on a professional high diver who performed sixteen dives from heights between 27-57 feet. For each dive, he executed a 3/4 turn and landed on his back supine on a three-foot thick mattress. Sagittal accelerations were recorded from a point on the sternum and on the forehead. From these data and a review of the literature it was determined that a healthy, adult male can voluntarily withstand posterior-anterior (P-A) chest decelerations of 50 Gs for pulse durations of less than 100 msec. A chest acceleration tolerance level of 60 Gs measured at the center of the chest is recommended for both P-A and A-P accelerations of 100 msec or less.

As a result of cadaver studies (20), serious questions have been raised as to the suitability of using sternal or spinal GSI acceleration measurements as indicators of possible thorax injury. Data from 18 unembalmed cadavers exposed to blunt thoracic impact indicated that G-levels/severity indicies at the sternum did not correlate with the Abbreviated Injury Scale (AIS) for injury. According to this analysis, a mild exposure (AIS <3) would have produced G-levels/severity indicies at the sternum much in excess of the recommended 60 G/1000 GSI tolerance levels, whereas the same values measured at the spine would have indicated an extremely severe exposure. Normalized chest deflections (i.e., the ratio of penetration to chest depth) correlate well with AIS ratings for blunt, midsternal, A-P thoracic impacts. Although the spinal severity index shows a moderate correlation with AIS rating (injury severity), this index is believed to be too strongly dependent on restraint to be useful.

Injuries to laboratory surrogates (unembalmed fresh cadavers) run on a sled with acceleration profiles calculated to simulate car impacts during actual front-end collisions (both direct and off-axis) were compared with actual injury statistics (21). One of the conclusions from these data was that the cadaver was a poor human surrogate for testing 3-point restraint belts in front end collisions because the cadavers, unlike living humans, showed evidence of severe thoracic and cervical injuries including multiple rib fractures, and fractures of the sternum, clavical and cervical vertebrae. The data for living humans was obtained from accident records.

The seated subjects in the NBDL experiments are restrained by an inverted V-type lap belt consisting of a crotch strap and two shoulder belts. Reference (9) notes that a shoulder belt tension load of 1300 lbs. is a more appropriate tolerance level than the blunt object impact severity discussed above. This 1300 lbs. limit compares to an allowable combined shoulder belt load of 1800 lbs. as determined from noninjurious human experiments. The disadvantage of using shoulder belt load as a tolerance measure is its sensitivity to geometry.

Making geometric assumptions based on a standard 50th percentile man,* an 1800-lb. combined load on the belts would result from a configuration in which the shoulder belt height was 65 cm above the seat pan and the sled acceleration was 29 Gs. Since non-NBDL human experiments have been run at G-levels in excess of 29 Gs in the same directions and with similar restraints as used at NBDL, it is predicted that this load on the torso in the -X direction is an acceptable tolerance level. For the +X direction where the load is transferred to the subject over a wide area of the seat back, a somewhat higher tolerance level would be acceptable. The acceptability of a tolerance level

*Excluding the mass of the legs and pelvis, the upper torso mass in the standard 50th percentile man is 40.4 kg with a center of gravity 47 cm above the seat pan.

of 30 Gs for both -X and +X experiments is reinforced by data in reference (23), which presents fully restrained human whole-body impact tolerance limits of 45 Gs in the -X direction based on a 250 G/sec onset rate and less than .1 second duration, and tolerance limits of 83 Gs in the +X direction for durations less than .04 sec.

Stapp and Taylor (24) exposed human volunteers to sled accelerations up to 24 ±Gy without voluntary tolerance limits being reached. In these experiments, head and neck motions were restrained by side panels. Therefore, for the ±Y direction, a torso tolerance level of 24 Gs in a well-restrained subject seems appropriate. With the head and neck unrestrained (as in the experiments at NBDL), neck motion would limit impact exposures to much less than the torso tolerance levels discussed previously (see Table 1). The impact tolerance levels recommended under reference (23) for the +Z and -Z directions are 20 Gs and 15 Gs respectively for acceleration pulse durations of less than .1 second.

Recommended human tolerance levels for whole-body impact accelerations in the -X, +X, +Z and -Z directions can be also presented as a function of acceleration pulse duration (25). Plots of these tolerance levels are presented in Figures 6 and 7. These figures indicate the limits of noninjurious voluntary human impact exposures. For the +X runs, the head was prevented from rotating backwards, thereby avoiding the whiplash-type of neck injury. These curves provide a tolerance level for the torso under restraint conditions similar to the experiments conducted at NBDL and, with the exception of +X exposures, these exposures are tolerated by the head and neck as well.

The torso tolerance levels discussed in this section are summarized in the table below. The peak sled accelerations, maximum allowable duration of the acceleration pulses, and the reference from which the tolerance level was obtained are indicated in this table. For the +Y and +X directions, the head and neck were constrained and the tolerance levels presented are for the torso only. Experiments at NBDL, with the possible exception of those in the ±Z direction, would be limited by the risk of injury to the head and neck rather than by torso limitations.

(1) Wholebody tolerance levels.

DIRECTION	ACCELERATION PEAK (SLED)	DURATION	REFERENCE
-X	40G	<.1 MSEC	25
-X	45G	<.1 MSEC	23
+X	34G	<.1 MSEC	25
+X	83G	<.04 MSEC	23
+Z	20G	<.1 MSEC	25
+Z	20G	<.1 MSEC	23
-Z	7G	<.1 MSEC	25
-Z	15G	<.1 MSEC	23
+Y	24G		24
-Y	23.1G	<.063 SEC	26

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TABLE 1. MOST SEVERE NBDL SLED PROFILE PARAMETERS

----- SLED ACCELERATION -----				ENDSTROKE				
DIRECTION (ANATOM.)	PEAK G	ONSET G/SEC	DURATION MSEC	VELOCITY M/SEC	NO. OF SUBJECTS*	RESTRAINT	REMARKS	EYEBALL MOTION
-X	15.9	1522	99	18	13	1	HOLD,NUCU	EBO
-X	15.6	484	93	18	17	1	HOLD,NUCU	EBO
-X	15.5	2129	25	4.2	11	1	HOSD,NUCU	EBO
+Y	7.2	693	78	6.3	20	2	HOLD,NUCU	EBR
+Y	7.2	. 162	66	6.5	29	2	LOLD,NUCU	EBR
+Y	11.3	1433	28	3.5	20	2	HOSD,NUCU	EBR
-X+Y	7.1	784	105	8.9	13	2	HOLD,NUCU	EBOR
-X+Y	9.1	899	87	9.0	5	2	HOLD,NUCU	EBOR
-X+Y	11.4	342	101	14.8	3	2	LOLD,NUCU	EBOR
-X+Y	9.2	235	74	9.0	19	2	LOLD,NUCU	EBOR
-X+Y	13	1987	27	3.8	16	2	HOSD,NUCU	EBOR
+Z	12.5	1070	91	11.9	8	3	HOLD,NUCU g-CHEST TO BACK HORIZONTAL SLED	EBF
+Z	10.5	853	101	11.6		3	HOLD,NUCU g-CHEST TO BACK	EBF

*NOT DIFFERENT SUBJECTS FOR EACH COMBINATION OF DIRECTION AND SLED ACCELERATION/VELOCITY PARAMETERS.

RESTRAINT:
1. SEATED UPRIGHT, SHOULDER AND LAP BELTS, INVERTED V PELVIC STRAP ATTACHED TO LAP BELT.
2. LIKE (1) WITH PADDED SIDEBOARD AGAINST RIGHT SHOULDER.
3. SUBJECT ON BACK WITH HEAD ON HEADREST, RESTRAINT LIKE (1).

REMARKS, EYEBALL MOTION:

EBO = EYEBALLS OUT
EBR = EYEBALLS RIGHT
EBOR = EYEBALLS OUT AND RIGHT
EBF = EYEBALLS FOOTWARD

HOLD = HIGH ONSET-LONG DURATION
HOSD = HIGH ONSET-SHORT DURATION
LOLD = LOW ONSET-LONG DURATION
NUCU = NECK UP, CHIN UP

GUIDELINES FOR SAFE HUMAN EXPERIMENTAL EXPOSURE TO IMPACT ACCELERATION

TABLE 2. FORCES AND TORQUES FOR HIGH EXPOSURE RUNS

RUN NO.	SUBJ	SLED ---- ACCELERATION ---- DIREC (ANAT.)	PEAK G	ONSET G/SEC	DUR. MS	ENDSTR. VELOC. M/SEC	MAX Fx N	MAX Fy N	MAX Fz N	MAX FR N	MAX Tx N-M	MAX Ty N-M	MAX Tz N-M	MAX TR N-M	MAX HIC /WINDOW WIDTH	MAX<20 HIC /WINDOW WIDTH
3990	131	-X	15.4	5[illegible]7	89	17.3	- 900	LESS THAN 100	- 680	1150	-	- 54	-	55	155 / 83	59 / 20
3983	134	-X	15.6	539	89	17.6	-1000	LESS THAN 100	- 750	1200	-	- 60	-	60+	187 / 84	71 / 20
4124	131	+Y	7.2	169	75	6.94	- 175	+ 390	- 305	475	- 33	LESS THAN ±10	+12.5	35	18 / 94	6 / 20
4126	134	+Y	7.1	167	75	6.90	- 260	+ 400	- 305	490	- 33	- 15	+ 11	35	21 / 91	7 / 20
4251	131	-X+Y	10.2	281	105	13.8	- 676	377	- 445	840	- 27	- 43	12	50	83 / 107	29 / 20
4307	134	-X+Y	11.4	337	101	14.9	- 675	363	- 620	900	- 31	- 45	12	50	97 / 87	35 / 20
4651	148	+Z	12.3	1135	93	11.8	260	130	800	850	17	- 35	-	38	54 / 71	26 / 20
4654	152	+Z	12.5	1070	91	11.9	400	280	700	750	- 22	+ 50	-	55	61 / 77	24 / 20
4742	D006	+Z	12.5	1070	91	11.9	220	65	900	900	- 7	+ 14 - 28	-	28	46 / 106	33 / 7

TABLE 3

PEAK RESULTANT VALUES OF HEAD KINEMATIC VARIABLES
DETERMINED FROM AVERAGE PROFILES

(AGARD CONFERENCE PROCEEDINGS-CP-253)

ACCEL. DIRECTION	ANGULAR ACCEL. RAD/SEC^2			ANGULAR VELOCITY RAD/SEC			LINEAR ACCELERATION M/SEC^2			PEAK SLED ACCELERATION
	HOLD	HOSD	LOLD	HOLD	HOSD	LOLD	HOLD	HOSD	LOLD	G-LEVEL
-X	1800	1600	1600	35	27	35	285	200	245	15
+Y	1030	860	900	26	20	24	130	78	123	7

TABLE 4

AVERAGE SHOULDER BELT AND SEAT PAN LOADS +Z DIRECTION
(ACCELERATION VECTOR CHEST TO BACK IN THESE EXPERIMENTS)

RUN NO.	SUBJECT NUMBER	PEAK SLED ACCEL - G	ONSET G/SEC	DURATION MSEC	ESV M/SEC	SEAT PAN LOAD N	AVERAGE BELT LOAD N
4651	148	12.3	1135	93	11.8	13600	635
4654	152	12.5	1070	91	11.9	9600	580
4742	HYB III 95%	12.5	1070	91	11.9	14700	345

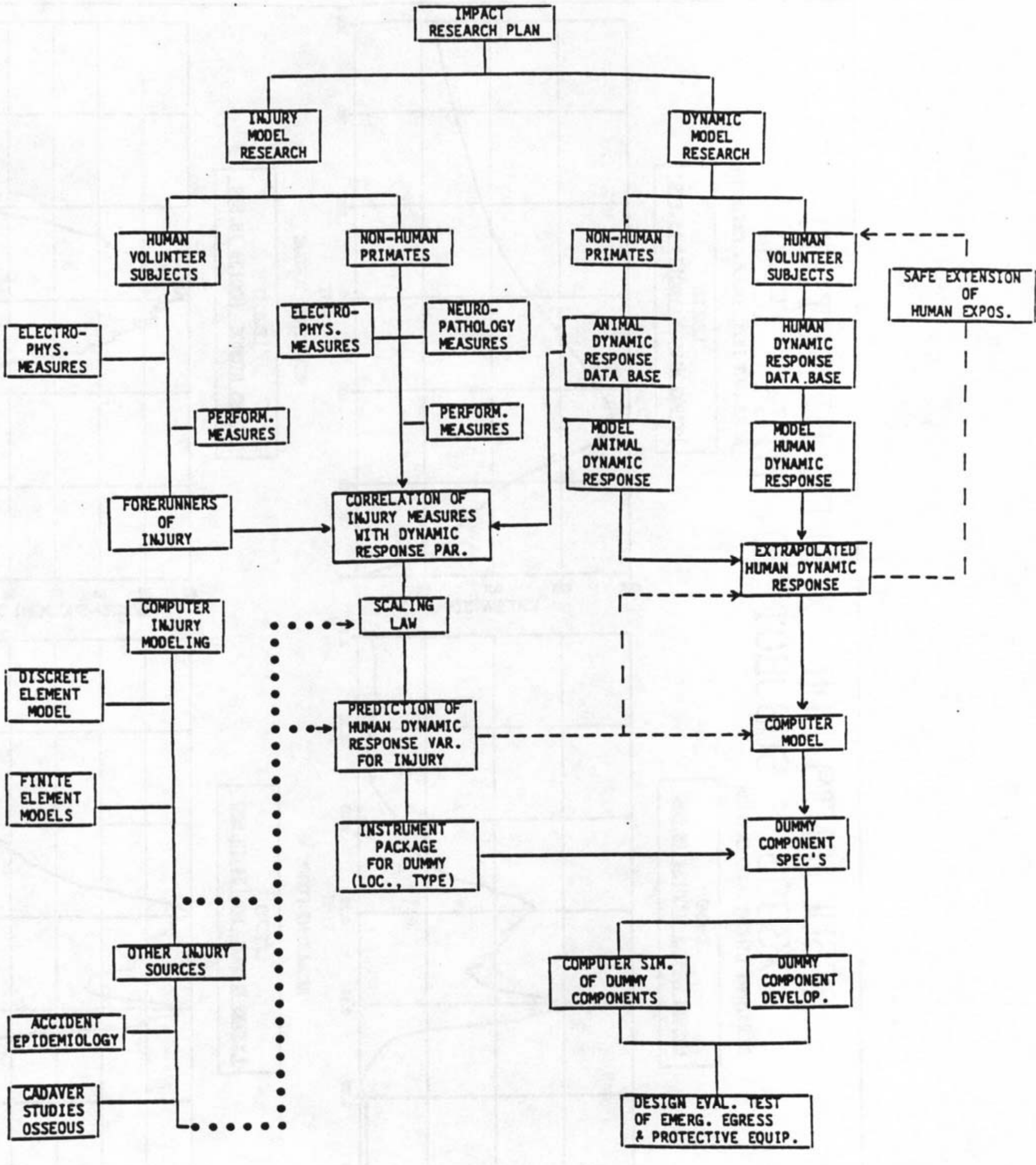

IMPACT RESEARCH PLAN
INJURY MODEL RESEARCH
DYNAMIC MODEL RESEARCH
HUMAN VOLUNTEER SUBJECTS
NON-HUMAN PRIMATES
NON-HUMAN PRIMATES
HUMAN VOLUNTEER SUBJECTS
SAFE EXTENSION OF HUMAN EXPOS.
ELECTRO-PHYS. MEASURES
ELECTRO-PHYS. MEASURES
NEURO-PATHOLOGY MEASURES
ANIMAL DYNAMIC RESPONSE DATA BASE
HUMAN DYNAMIC RESPONSE DATA BASE
PERFORM. MEASURES
PERFORM. MEASURES
MODEL ANIMAL DYNAMIC RESPONSE
MODEL HUMAN DYNAMIC RESPONSE
FORERUNNERS OF INJURY
CORRELATION OF INJURY MEASURES WITH DYNAMIC RESPONSE PAR.
EXTRAPOLATED HUMAN DYNAMIC RESPONSE
COMPUTER INJURY MODELING
SCALING LAW
DISCRETE ELEMENT MODEL
PREDICTION OF HUMAN DYNAMIC RESPONSE VAR. FOR INJURY
COMPUTER MODEL
FINITE ELEMENT MODELS
INSTRUMENT PACKAGE FOR DUMMY (LOC., TYPE)
DUMMY COMPONENT SPEC'S
OTHER INJURY SOURCES
COMPUTER SIM. OF DUMMY COMPONENTS
DUMMY COMPONENT DEVELOP.
ACCIDENT EPIDEMIOLOGY
CADAVER STUDIES OSSEOUS
DESIGN EVAL. TEST OF EMERG. EGRESS & PROTECTIVE EQUIP.

Fig 1

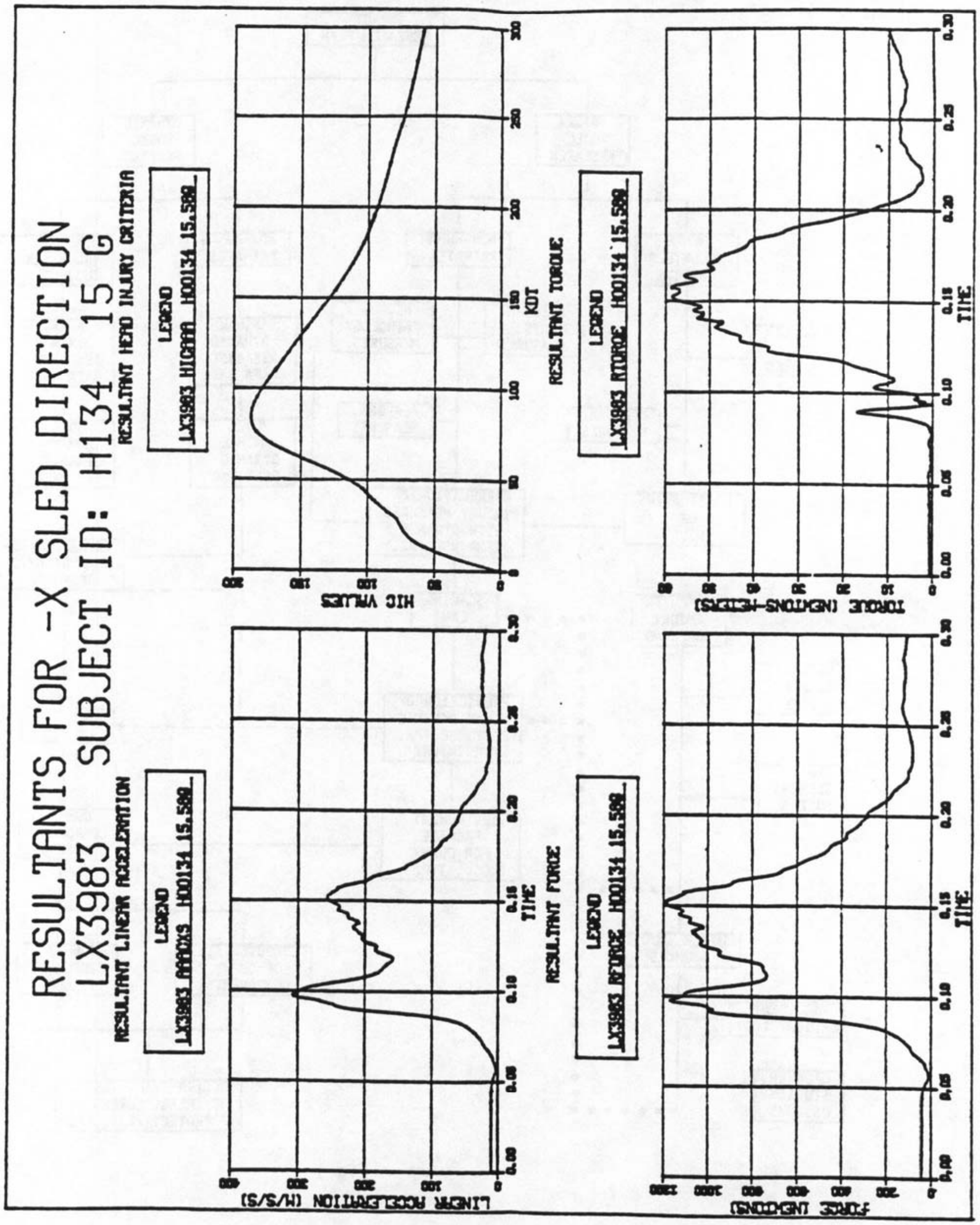
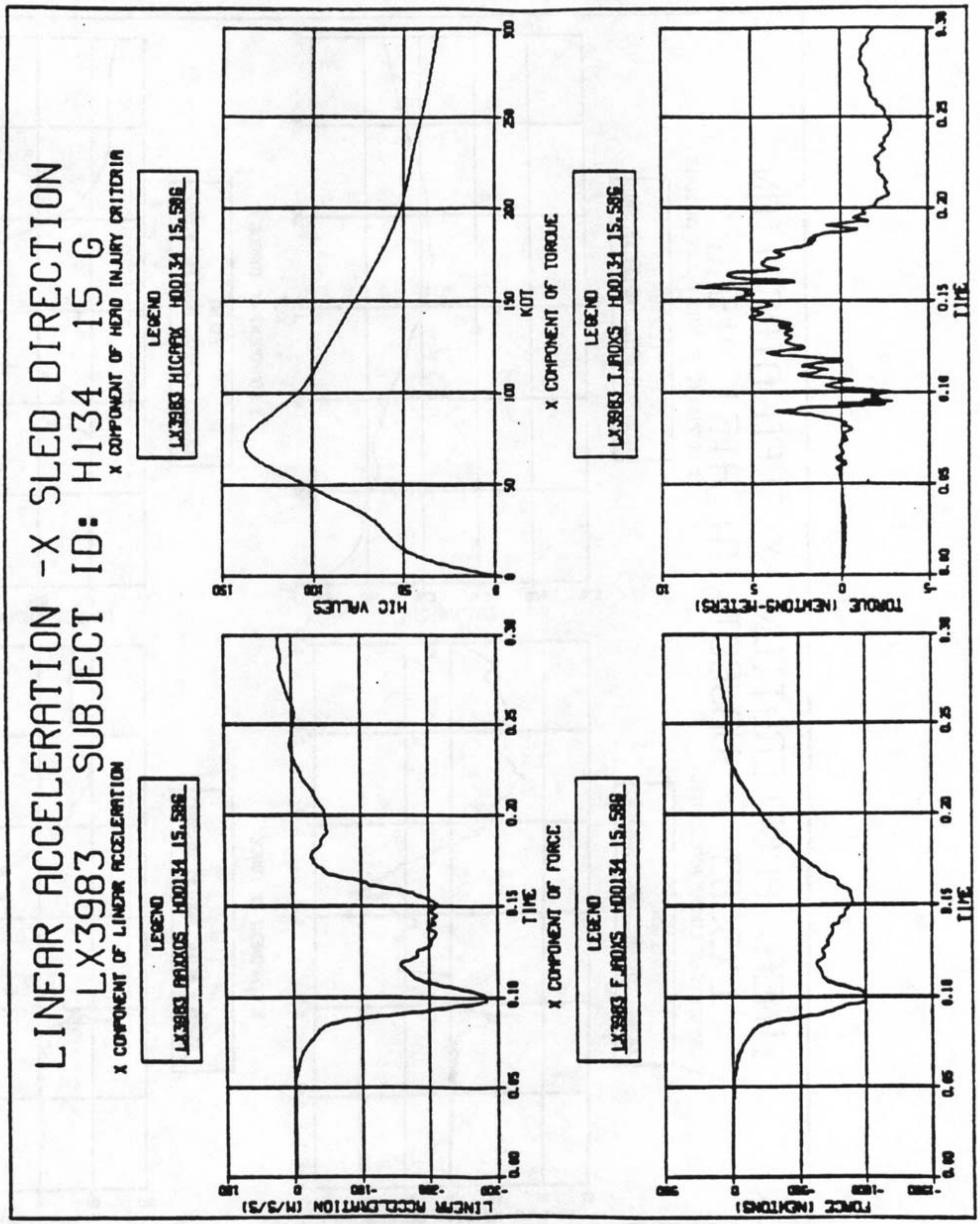

RESULTANTS FOR -X SLED DIRECTION
LX3983 SUBJECT ID: H134 15 G
RESULTANT LINEAR ACCELERATION
LEGEND
LINEAR ACCELERATION (M/S/S)
TIME
RESULTANT HEAD INJURY CRITERIA
LEGEND
HIC VALUES
KDT
RESULTANT FORCE
LEGEND
FORCE (NEWTONS)
TIME
RESULTANT TORQUE
LEGEND
TORQUE (NEWTONS-METERS)
TIME

Fig. 2



LINEAR ACCELERATION -X SLED DIRECTION
LX3983 SUBJECT ID: H134 15 G
X COMPONENT OF LINEAR ACCELERATION
X COMPONENT OF HEAD INJURY CRITERIA
X COMPONENT OF FORCE
X COMPONENT OF TORQUE
LEGEND
LINEAR ACCELERATION (M/S/S)
HIC VALUES
FORCE (NEWTONS)
TORQUE (NEWTONS-METERS)
TIME
KDT

Fig. 3

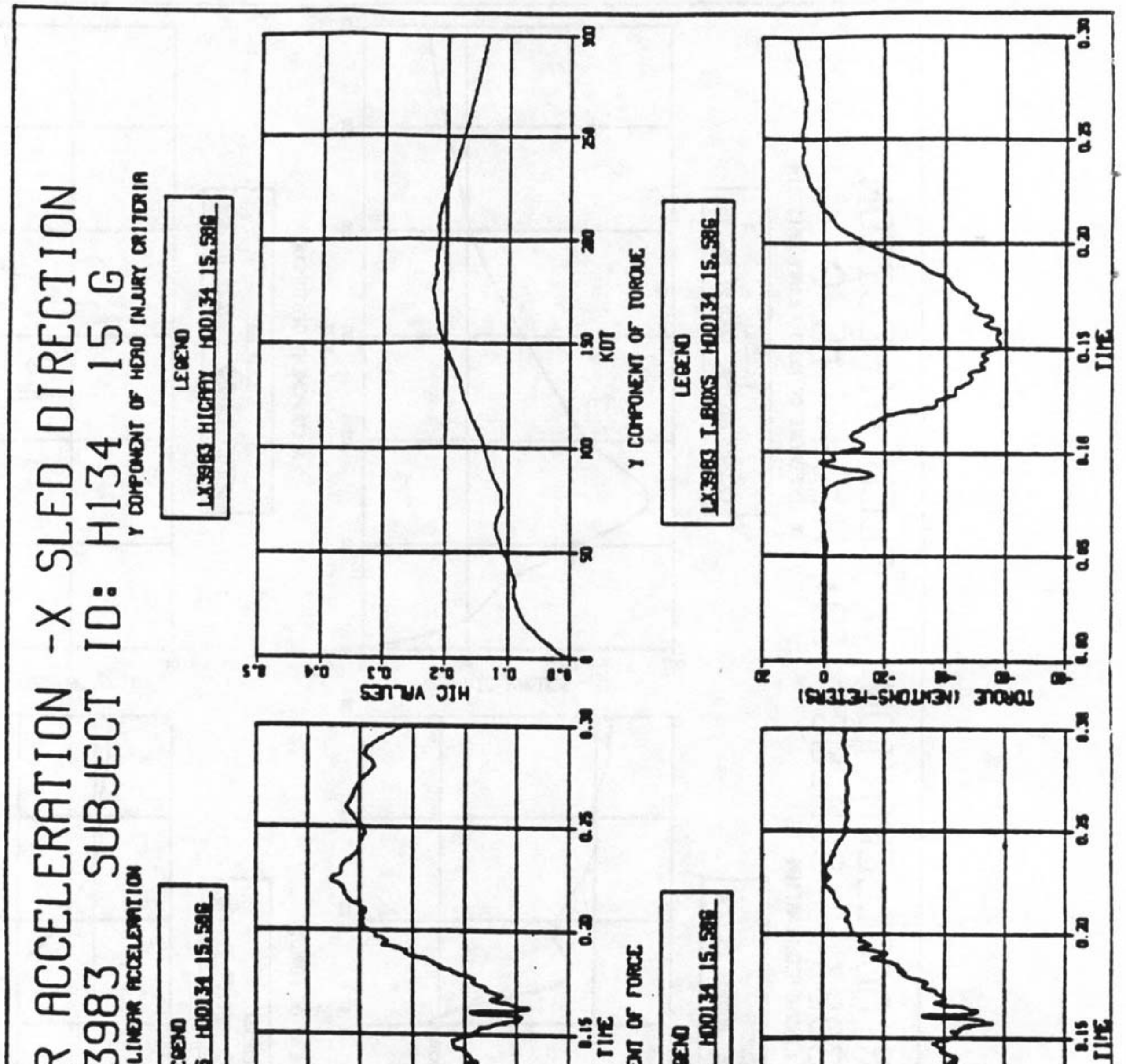

LINEAR ACCELERATION -X SLED DIRECTION
LX3983 SUBJECT ID: H134 15 G
Y COMPONENT OF LINEAR ACCELERATION
LEGEND
LINEAR ACCELERATION (M/S/S)
TIME
Y COMPONENT OF HEAD INJURY CRITERIA
LEGEND
HIC VALUES
KDT
Y COMPONENT OF FORCE
LEGEND
FORCE (NEWTONS)
TIME
Y COMPONENT OF TORQUE
LEGEND
TORQUE (NEWTONS-METERS)
TIME

Fig. 4

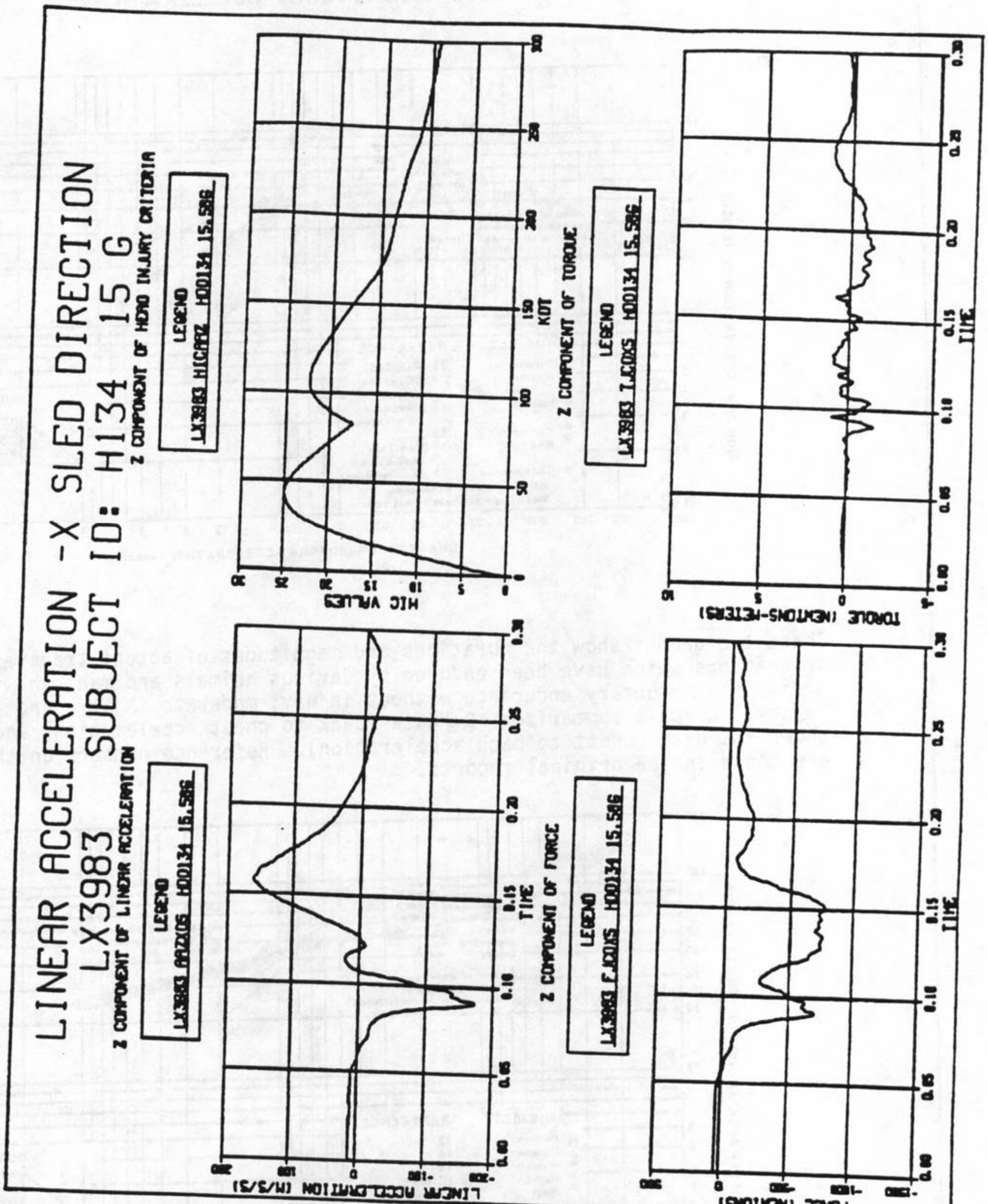
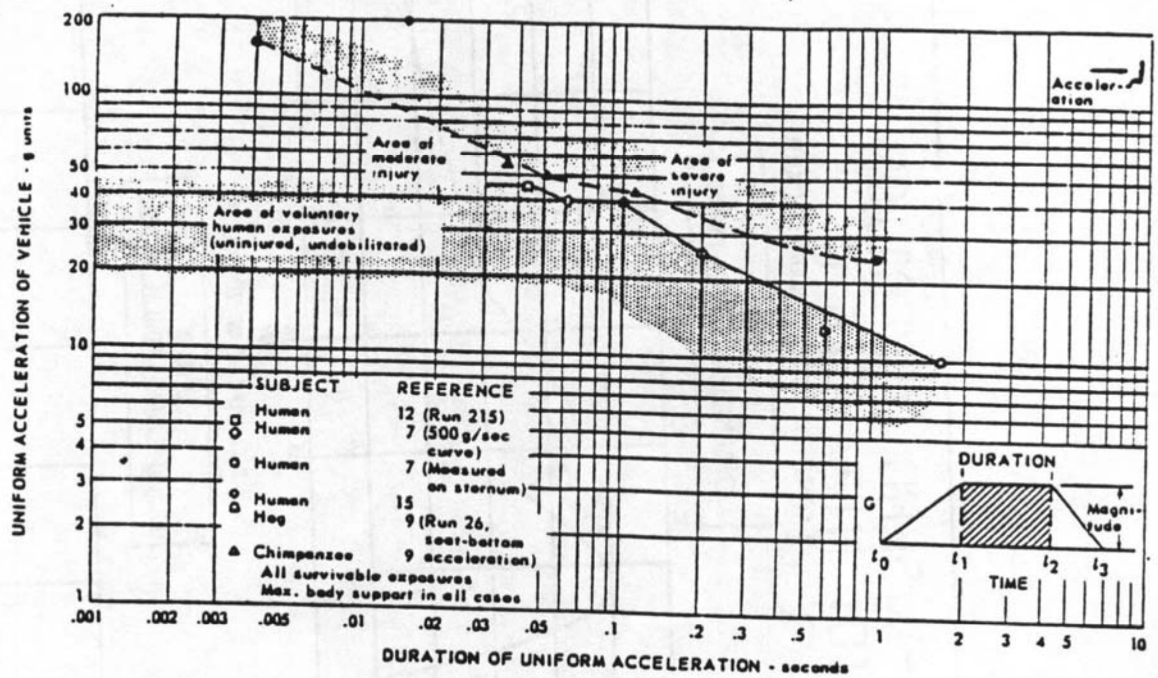

LINEAR ACCELERATION -X SLED DIRECTION
LX3983 SUBJECT ID: H134 15 G
Z COMPONENT OF LINEAR ACCELERATION
LEGEND
LINEAR ACCELERATION (M/S/S)
TIME
Z COMPONENT OF HEAD INJURY CRITERIA
HIC VALUES
KDT
Z COMPONENT OF FORCE
FORCE (NEWTONS)
Z COMPONENT OF TORQUE
TORQUE (NEWTONS-METERS)

Fig. 5

ABRUPT TRANSVERSE DECELERATIONS

a.



These two graphs show the durations and magnitudes of abrupt transverse decelerations which have been endured by various animals and man, showing areas of: voluntary endurance without injury; moderate injury; and severe injury. Graph a summarizes $-G_x$ data (back to chest acceleration) and b shows $+G_x$ data (chest to back acceleration). Reference numbers on the graphs are those in the original reports.

b.

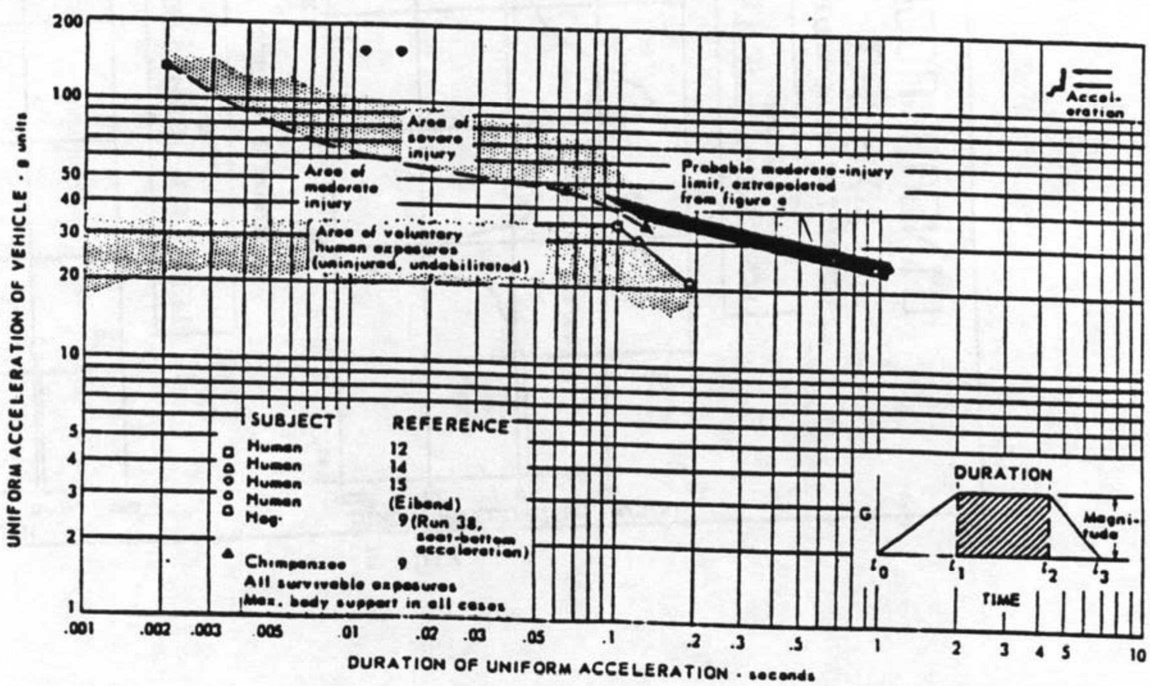
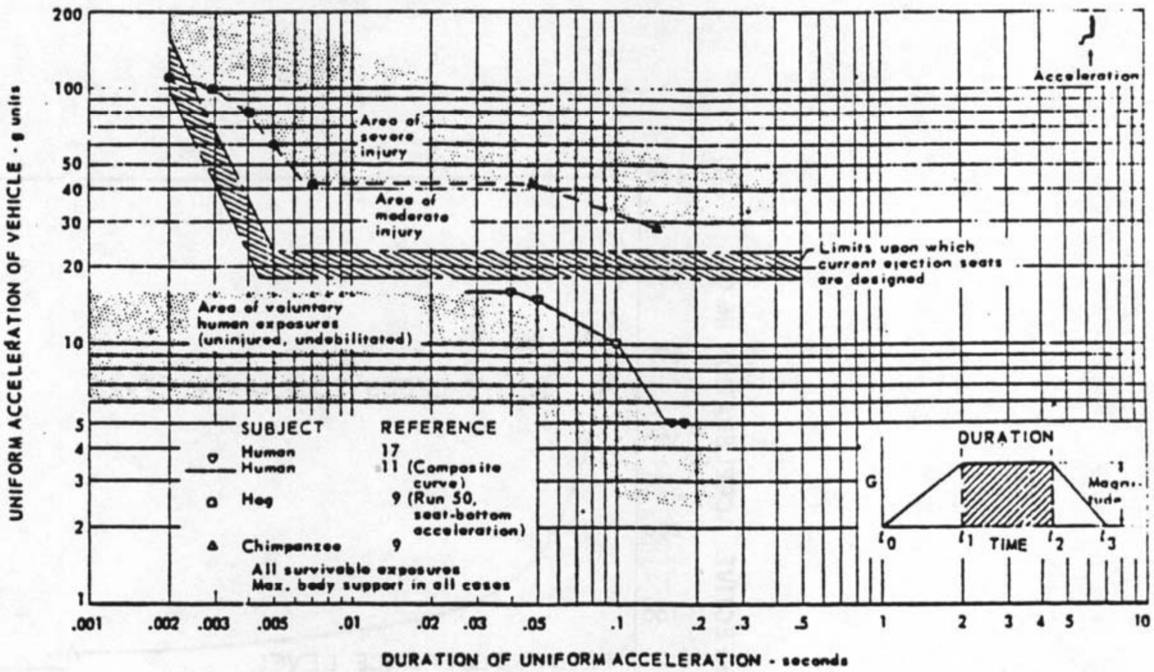


Fig. 6

ABRUPT LONGITUDINAL DECELERATIONS

a.



These two graphs show the durations and magnitudes of abrupt deceleration in the G (longitudinal) directions which have been endured by various animals and man, showing areas of voluntary endurance without injury, moderate injury, and severe injury marked by shading. Graph a shows data of $+G_z$ acceleration (headward), and b shows data for $-G_z$ acceleration (tailward). Reference numbers on the graphs are those in the original reports.

b.

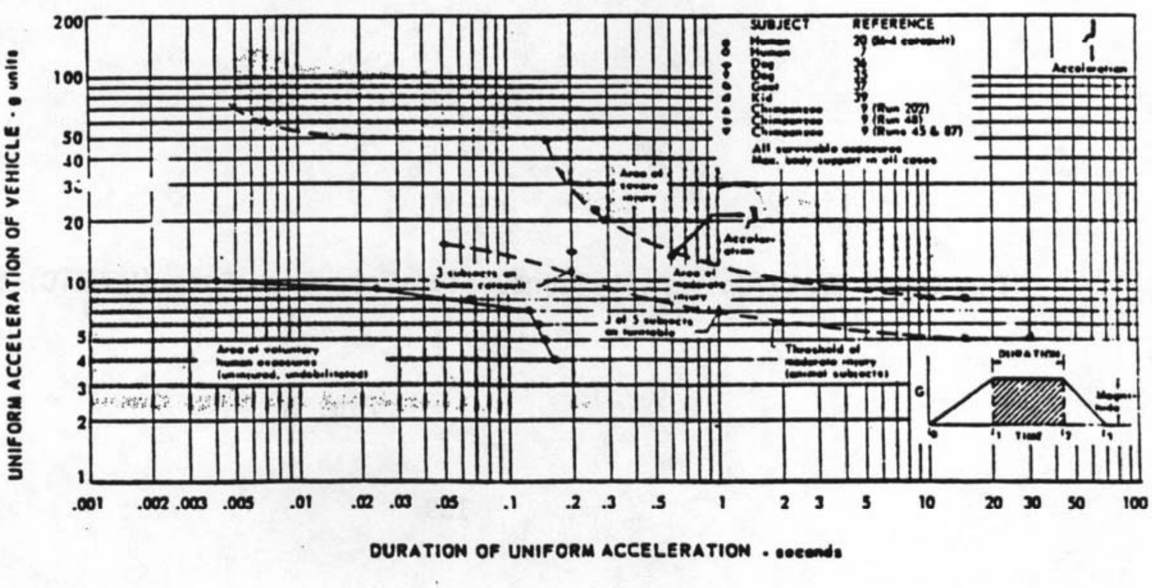
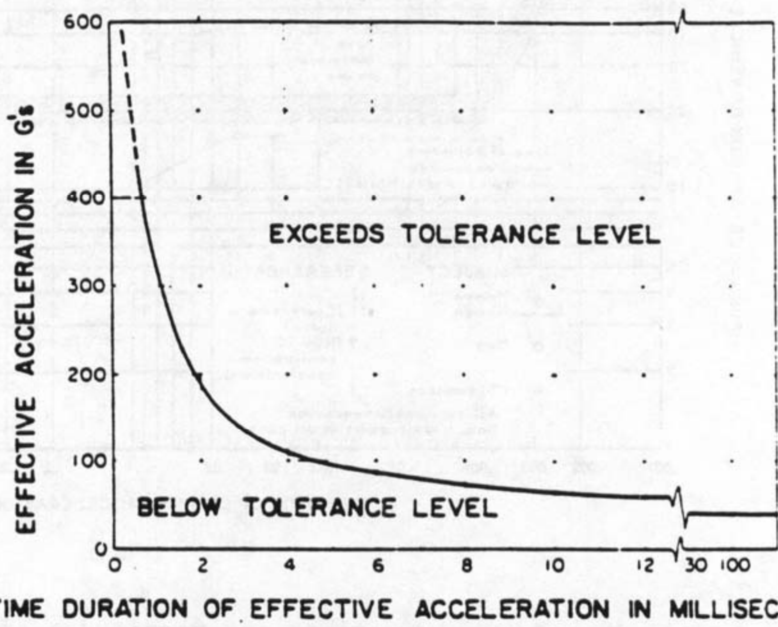


Fig. 7



IMPACT TOLERANCE FOR THE HUMAN BRAIN IN FOREHEAD IMPACTS AGAINST PLANE, UNYIELDING SURFACES

The Wayne State University Concussion Tolerance Cure (WSUCTC) taken from ref. (9)

Fig. 8

GUIDELINES FOR SAFE HUMAN EXPERIMENTAL EXPOSURE TO IMPACT ACCELERATION

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GUIDELINES FOR SAFE HUMAN EXPERIMENTAL EXPOSURE TO IMPACT ACCELERATION

[illegible] [illegible] of the Kinematics of Whiplash, [illegible] [illegible] Stapp Car Crash Conference, 1967.

[illegible] Mechanics of Head Injury, Goldsmith, W., in: Biomechanics [illegible] and Orthopedics, [illegible] et al., [illegible] 1972.

[illegible] Transient Intracranial [illegible] Pressure During [illegible] in Human Subjects, [illegible] et al., in Models and Analogues for the Evaluation of Human [illegible] Response, Performance and [illegible] Conference Proceedings [illegible] 1970.

[illegible] Protection of the Head and Neck in Sports, [illegible] and Patrick, L.M., JAMA, 182:509-512, 1962.

DISCUSSION

PAPER: Safe Human Experimental Exposure to Impact

SPEAKER: Marc S. Weiss, Naval Biodynamics Lab

Q: Lloyd Thompson, McGill University

Several years ago there was a STAPP meeting in New Orleans, and I was privileged, along with many others, to visit your laboratory. At that time, as I understood it, your volunteers received special preparation for strengthening neck muscles.

A: Weiss

Not true.

Q: Also, did they decide when the blast would come?

A: No.

Q: Well, maybe I am talking about a different facility?

A: The volunteer hazard controller has stimulating electrodes on one hand and on the other hand he has a switch, which is taped to his hand. He's got to keep the button depressed. If he releases that button anytime up to the time of fire the experiment is aborted. Now, it is again a matter of historical interest that that has never happened. We have had,on occasion, two or three times, subjects faint on the sled; due to a combination of stress of the experiment, anticipatory anxiety, or tightness of the restraint. They have fainted and that button was still locked tight. It was a medical officer who aborted that experiment. You have to realize these are young military guys and you don't quit. We have had only one volunteer in the history of the laboratory who dropped out of the program. He developed an aphobic response to the sled, much to his total bewilderment and amazement. He was more at a loss to understand what was going on than anyone else.

Q: Albert King, Wayne State University

Can you comment on the repeated exposure of these subjects to the impacts?

A: I'm not sure what your asking specifically. These subjects experience numerous exposures. Many of them in several vector directions. I'm not sure what you mean?

Q: I'm wondering about possible low level brain injury due to repeated impact?

A: Okay. I'm glad you raise that question. We do have, as an integral part of our program, what we call our long term follow-up. That's not the official name, but that's what we call it. We bring these guys back every three years and they get a complete work-up. First of all, its important to note, the physical screening these guys get for this program is far more extensive than the screening that they get to come into the military. The medical reasons for rejection are basically bone or orthoapedically related. Either they have dentition problems, their teeth have to be in pretty fair shape, so we can fit them. The other problem is spinal abnormalities, which don't keep them out of the service but keep them out of our experiment. Scoliosis is the most common kind of problem that we don't want to get involved with. We have not had any problems, What's the longest we go back now, do you know Bill?

A: 10 years.

A: Okay. For ten to fifteen years now we have been tracking some of these people and the results have been negative. There's no detectably significant or medically interesting problem that can in any way be related to their exposures on the sled.

Q: Guy Nusholtz, Chrysler

What type of accelerometers have you been using?

A: Bill can answer that better than I.